Individual Muscle Forces during Sit to Stand Transfer

Elena J. Caruthers

Prepared for the 2014 Hayes Graduate Research Forum at The Ohio State University

April 15, 2014[1]

Abstract

Due to weakened muscles or diseased joints, more than 2 million Americans over the age of 64 have difficulty accomplishing a sit-to-stand (STS) transfer independently. Previous studies have examined joint torques and muscle activations during STS by using motion capture or rigid body models. However, individual muscle forces during STS have yet to be investigated, and such knowledge will potentially inform rehabilitation strategies for patients with weakened muscles to improve performance with STS transfer. The first step toward accomplishing this goal was to examine individual muscle forces as well as inter-limb differences in muscle forces during STS transfer in a young, healthy population. Subject-specific simulations were created for each subject's STS trial with a custom three-dimensional musculoskeletal model. Static optimization was implemented to estimate individual muscle forces. We found that vastus lateralis generated the largest force, reaching its peak value after maximum hip flexion occurred, while the medial gastrocnemius generated the smallest force out of all the muscles examined throughout STS once maximum hip flexion was reached. Inter-limb differences, quantified as a percent difference, showed high variability between subjects as the standard deviation values were over 100% for some of the muscles examined across the phases of STS. Understanding individual muscle forces as well as symmetry of muscle forces between legs during STS transfer in healthy subjects is the first step to analyzing muscle function and weakness in patients with pathologic conditions such

[1] The co-authors of this paper are Julie A. Thompson, Ajit M.W. Chaudhari, Laura C. Schmitt, Thomas M. Best, Katherine Saul, and Robert A. Siston.

as osteoarthritis and may potentially inform rehabilitation strategies that could improve these patients' functional performance with this task.

Introduction

Being able to rise and stand from a seated position is a basic, yet biomechanically demanding, activity of daily living (Hughes and Schenkman, 1996). However, due to weakened muscles or diseased joints, more than 6% of community-dwelling older adults (Leon et al., 1990) and over 60% of nursing home residents (Mehr and Fries, 1993) have difficulty accomplishing this sit-to-stand (STS) transfer independently, which can significantly limit mobility and independence (Wretenberg and Arborelius, 1994). STS transfer is often used as a test to assess functional performance in older people with lower limb strength or patients with conditions such as arthritis and proximal myopathy (Lord et al., 2002). A time greater than 1.85 ± .28 seconds to complete the STS transfer under natural conditions usually indicates poor functional mobility (Hanke et al., 1995).

To complete the STS transfer successfully, sufficient strength and power must be generated by the leg muscles to develop adequate joint torques to allow individuals to raise their center of mass (Hanke et al., 1995; Schultz et al., 1992). Several studies have divided the STS transfer into different phases, including: forward leaning phase, momentum transfer phase, and extension phase (Mak et al., 2003; Schenkman et al., 1990; Schenkman et al., 1996; Schultz et al., 1992; Su et al., 1998). EMGs have shown that the quadriceps (vasti and rectus femoris), biceps femoris, gluteus maximus, and tibialis anterior are active during both the forward leaning phase and momentum transfer phase, producing a greater joint torque at the knee, followed in magnitude by the hip, and then the ankle (Doorenbosch et al., 1994; Mak et al., 2003; Munton et al., 1984; Rodrigues-de-Paula Goulart and Valls-Sole, 1999; Yoshioka et al., 2012). At the end of extension phase, the hamstrings, soleus, and gastrocnemius are active to produce the highest joint torque at the ankle, followed by magnitude in the hip, and then the knee (Doorenbosch et al., 1994; Mak et al., 2003; Munton et al., 1984; Rodrigues-de-Paula Goulart and Valls-Sole, 1999; Yoshioka et al., 2012).

However, the function of individual muscles cannot be clearly determined from joint torques and muscle activation alone due to the complex dynamics of the human body (Delp et al., 2007). This complexity, in part, derives from the fact that muscles can accelerate joints that they do not span and body segments to which they do not attach (Delp et al., 2007). Understanding individual muscle function potentially informs rehabilitation strategies to improve patients' functional performance when completing tasks such as STS transfer. Previous studies have attempted to implement biomechanical rigid-body models to examine STS transfer with a subject's body segments configured in a variety of initial seating positions (Alexander et al., 2001; Doorenbosch et al., 1994; Scarborough et al., 2007; Schultz et al., 1992; Su et al., 1998). A majority of these models were driven by net reaction forces and joint torques calculated using equations of equilibrium, not by muscle forces. Limited work has explored the use of a muscle-driven model to determine the minimum muscle force required to complete STS transfer successfully (Ellis et al., 1984; Yoshioka et al., 2012).

In order to be able to inform rehabilitation treatments that could potentially improve patients' functional performance in the STS transfer, individual muscle forces and contributions to the phases of STS need to be considered when evaluating the relationship between muscle function and weakness with patient performance during STS transfer. The purpose of this study was to 1) examine individual muscle forces in STS transfer as well as 2) identify inter-limb differences in maximum muscle forces per phase of STS transfer in a young, healthy population.

We hypothesized that for a healthy population to perform STS transfer, 1) the quadriceps, hamstrings, and gluteus maximus would generate the largest forces of all the muscles examined and 2) that there would be no statistically significant inter-limb differences in the forces produced by these muscles across all phases of the STS transfer. We generated simulations using a custom three-dimensional musculoskeletal model to analyze the kinematics of STS transfer of young, healthy subjects to serve as a baseline for when we examine muscle function during STS transfer in the future in pathological populations.

Methods

Experimental Data

Seven healthy subjects (5 male and 2 female, Age: 22.7 ± 2.9 years, Mass: 78.2 ± 10.8 kg, Height: 1.77 ± 0.06 m) provided written informed consent in accordance with the Institutional Review Board of The Ohio State University to participate in this study. Subjects completed three trials of the STS transfer from a 55.2 cm chair while motion data was collected at 150 Hz using an 8-camera Vicon MX-F40 system; the marker sets used for the lower limb were based off of the Point-Cluster Technique (PCT) (Andriacchi et al., 1998) while the upper extremities were tracked with markers placed on their bony landmarks. Subjects began each trial by sitting in a chair with their arms crossed against their chest. While keeping their arms crossed against their chest, the subjects were asked to rise from the chair, pause for 2 seconds, and then return to sitting in the chair. Ground reaction forces were obtained from two force plates sampled at 600 Hz (Bertec, Columbus, OH), one placed under each of the subject's feet; no part of the chair touched the force plates. Muscle activation patterns from the gluteus maximus, gluteus medius, rectus femoris, vastus lateralis, biceps femoris, tibialis anterior, medial gastrocnemius, and soleus of both legs were measured with 16-channel surface EMG (Noraxon, Scottsdale, AZ) and sampled at 1500 Hz. EMG data were high-pass filtered at 10 Hz, rectified, and RMS smoothed with a 20 ms window.

Musculoskeletal Modeling

When analyzing STS transfer, we wanted to use a three-dimensional musculoskeletal model that would allow us to analyze the lower extremities but also had flexibility in the back and included arms for completeness. Since, at the time, no previously developed model fit all of these criteria, we created a three-dimensional custom musculoskeletal model, the Full Body Model 2013, by incorporating the following models: Lower Limb Model 2010 (Arnold et al., 2010), Musculoskeletal Model of the Lumbar Spine (Christophy et al., 2012), Upper Extremity Model 2013 (Holzbaur et al., 2005; Holzbaur et al., 2007) and Head and Neck Musculoskeletal Biomechanics Model (Vasavada et al., 1998) (Figure 1). This custom model has 46 degrees of freedom with 194 Hill-type muscle-tendon actuators. To ensure that adding a new torso and arms did not greatly affect the dynamics of the lower limb, our model’s muscle properties (i.e. moment arms) as well as kinetic, kinematic, and static optimization results were compared to those from the Gait 2392 model, a generic musculoskeletal model frequently used in OpenSim that focuses on the lower extremity musculature, for a single gait trial from a representative subject. These results compared favorably considering that the Lower Limb Model 2010 incorporated into the Full Body Model 2013 has longer fiber lengths for these muscles and more accurate force generation of the lower limb than the Gait 2392 model.

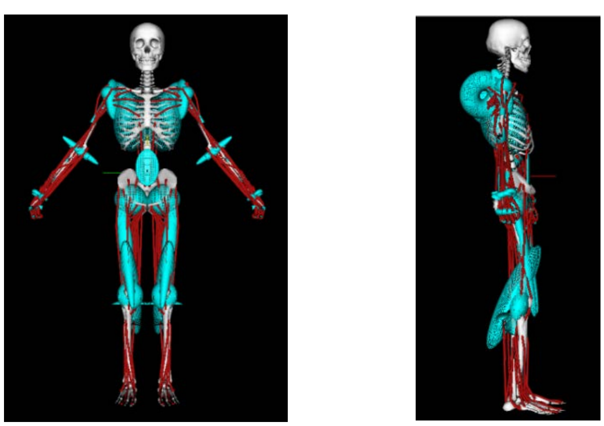

Figure 1: Anterior and lateral view of the Full Body Model 2013, which incorporates the following models: Lower Limb Model 2010, Musculoskeletal Model of the Lumbar Spine, Upper Extremity Model 2013, and Head and Neck Musculoskeletal Biomechanics Model. The blue spheres and cylinders shown with the model are wrapping surfaces used to define muscle-tendon paths so that they are more realistic to the human body, rather than just using point-to-point representations of muscles.

Simulations

Using OpenSim 3.1, each subject's model was created by scaling the Full Body Model 2013 to match their anthropometric data. The dimensions of each body segment in the model were scaled based on relative distances between pairs of markers obtained from motion capture during the static calibration trial and the corresponding virtual marker locations in the model so that the RMS marker error was no more than 3 cm. To simplify the upper extremity portion of the model and improve computational time, the upper extremities were locked into their initial position and had their muscles removed. The next step was to solve an inverse kinematics problem with a least-squares approach to minimize the difference between the experimental marker locations and the model's virtual marker locations. The inverse dynamics tools was then implemented on the subject's model for each STS trial to determine the net torques at each joint that drove their STS transfer motion (Delp et al., 2007). These net joint torques were further resolved into individual muscles forces by using the static optimization tool (STO); these forces are resolved by minimizing the sum of squared muscle activations at each instant of time of the given motion (Crowninshield, 1978; Delp et al., 2007). The results of STO were considered acceptable if the constraint violation given for each time step had a value between E-13 and E-12. To ensure that the simulated and experimental activation patterns were in agreement, the simulated muscle activations from STO were compared to the subject's experimental EMG after having normalized them by peak value of the simulated muscle activation (Figure 2).

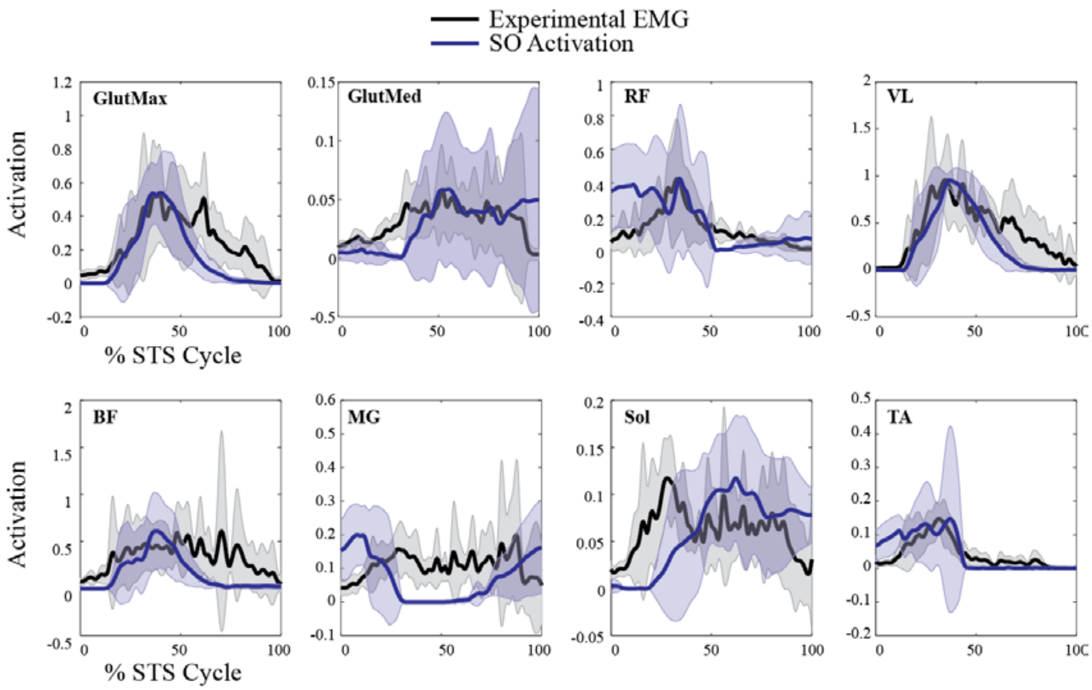

Figure 2: Experimental EMG (black) and simulated muscle activations (blue) in the dominant leg averaged over 7 subjects with shaded areas showing one standard deviation. The peak value of the experimental EMG is normalized to the peak value of the simulated muscle activation. (GlutMax = gluteus maximus, GlutMed = gluteus medius, RF = rectus femoris, VL = vastus lateralis, BFlh = biceps femoris long head, MG = medial head of gastrocnemius, Sol = soleus, and TA = tibialis anterior).

Analysis

Based on other studies, we chose to divide the STS transfer into the following phases: forward leaning phase, momentum transfer phase, and extension phase (Mak et al., 2003; Schenkman et al., 1990; Schenkman et al., 1996; Schultz et al., 1992; Su et al., 1998). Because there was no force plate data from underneath the chair, these phases were based off of kinematic values published in a previous study (Schenkman et al., 1990). The forward leaning phase (Phase 1) begins when lumbar extension increases by .5 degrees from its initial value when the subject is at rest. The momentum transfer phase (Phase 2) begins when the trunk and hip flexion angle reach their maximum value, as determined by the inverse kinematic analysis. The momentum transfer phase ends and the extension phase (Phase 3) begins when the ankle reaches its maximum dorsiflexion value (Schenkman et al., 1990). We chose to focus on specific muscles and muscle groups during these phases of STS transfer, including the gluteal (gluteus maximus, gluteus medius), quadriceps (rectus femoris, vastus lateralis, vastus intermedius, vastus medialis), hamstrings (semitendinosus, semimembranosus, biceps femoris), sartorius, gastrocnemius, soleus, and tibialis anterior muscles. After estimating muscle forces that were generated for each of the subjects' STS trials, we determined that each subject's data was sufficiently consistent enough between trials for us to choose one trial per subject for analysis. We then averaged the individual muscles forces across subjects and examined them across the three phases of STS transfer.

A three-way repeated measures analysis of variance (ANOVA) was performed to examine which muscles generated the greatest force across the phases of STS transfer. We examined the individual muscles, each phase of STS transfer, and leg (dominant and non-dominant) as main effects as well as the interaction effects between muscle, phase, and leg. The dominant leg was assigned as the one that generated a greater ground reaction force during the entire STS transfer. Tukey post-hoc pairwise comparisons were further used, as appropriate. These statistical tests were performed in Minitab® Statistical Software (Minitab Inc, State College, PA) and the level of significance was set to $\alpha = .05$.

Because subjects had the tendency to lean sideways or have valgus knee positioning during the STS transfer, inter-limb differences in muscle forces were also examined. Inter-limb differences for each phase of STS were quantified as the percent difference in maximum muscle forces between the dominant and non-dominant legs and averaged across subjects. A paired t-test was implemented using Minitab® to determine if any of these inter-limb differences were statistically significant ($\alpha = .05$).

Results

Joint Moments

The average hip and knee flexion moments across the seven subjects reach their maximum values of 64.82 ± 24.21 Nm and 75.11 ± 22.02 Nm, respectively, at the end of Phase 2 before the maximum ankle dorsiflexion is reached (Figure 3a). As the hip and knee joints goes into extension in Phase 3, the average hip and knee flexion moments decrease while the average ankle plantarflexion moment increases. The average ankle plantarflexion moment eventually reaches its maximum value of 24.05 ± 8.20 Nm when a standing position is achieved at the end of Phase 3 (Figure 3a).

Individual Muscle Forces

As the torso begins to bend forward at the beginning of Phase 1, the rectus femoris (RF) generates the greatest force of all the muscles examined, starting at 277.04 ± 187.22 N. As trunk flexion continues to increase, the vasti muscles as well as the gluteus maximus (GlutMax) and biceps femoris long head (BFlh) muscles begin to steadily increase in force generation. These muscles eventually increase to their peak values (vastus lateralis = 1,695.10 ± 169.64 N, vastus medialis = 1,028.90 ± 239.44 N, vastus intermedius = 700.91 ± 300.60 N, gluteus maximus = 952.08 ± 291.61 N, biceps femoris long head = 411.04 ± 284.69 N) at the end of Phase 2 before maximum ankle dorsiflexion is reached. Right before Phase 2 ends, the forces in the quadriceps (vasti and rectus femoris), gluteus maximus, biceps femoris long head muscles decrease while the force generated by the soleus increases. The soleus force continues to increase in Phase 3, reaching its peak value of 363.87 ± 238.68 N, as a standing position is attained. The medial head of the gastrocnemius (MG) increases in force around 70% of STS transfer while the quadriceps and gluteus maximus muscle forces continue to decrease (Figure 3b).

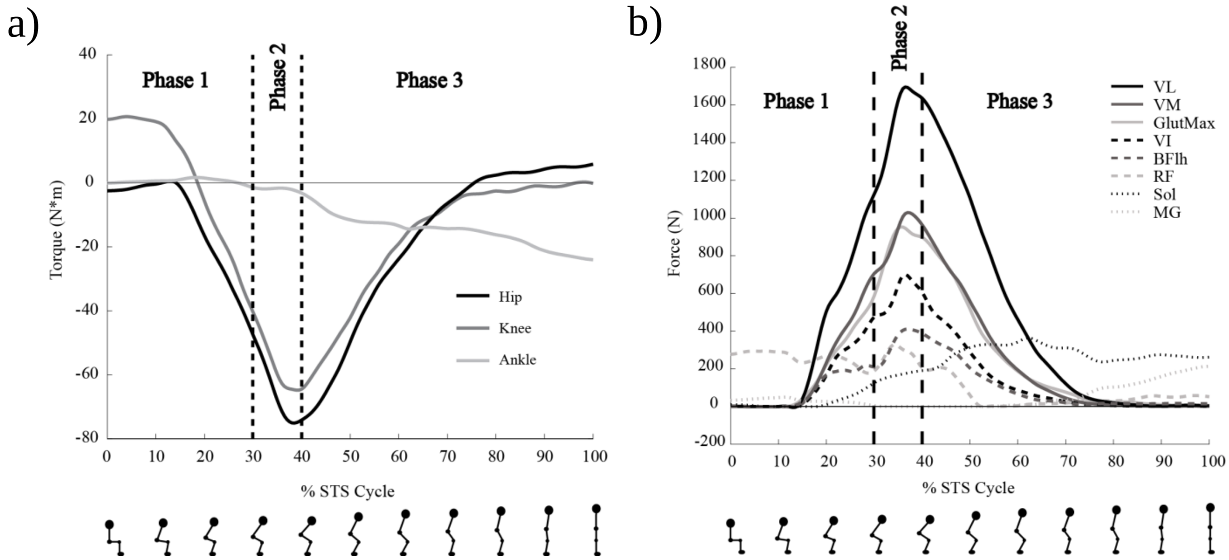

Figure 3: a) Average joint flexion torques for the dominant leg of seven subjects. Negative values indicate hip flexion, knee flexion, and ankle dorsiflexion moments. Hip and knee flexion torques reach their maximum value at the end of Phase 2 while ankle dorsiflexion torque reaches its maximum at the end of Phase 3. **b)** Average muscle forces for the dominant leg of seven subjects (VL = vastus lateralis, VM = vastus medialis, GlutMax = gluteus maximus, VI = vastus intermedius, BFlh = biceps femoris long head, Sol = soleus, and MG = medial head of gastrocnemius).

Different muscles produced significantly different maximum forces during the STS transfer ($p < .001$). The phases of STS transfer ($p < .001$) as well as the interaction between leg and muscle ($p = .013$) had significant effects on the maximum muscle forces. The phase and muscle interaction was also significant ($p < .001$), indicating that different muscles produce significantly different forces in the dominant and non-dominant leg across different phases of STS transfer. Of all the muscles examined, the vastus lateralis (VL) generates the largest force during STS transfer across all three phases (Tables 1 and 2). The results also showed that the force generated by the medial gastrocnemius (MG) was significantly different for Phases 2 and 3, generating the smallest force of all the muscles examined. Among the other muscles, there was high variability in force generation as the mean maximum muscle force values had large standard deviation values across phases (Table 2).

Table 1: Overall mean maximum muscle forces during STS transfer. The values represent the mean and standard deviation of the maximum muscle forces across the seven subjects for the entire STS transfer, including both dominant and non-dominant legs. Means that do not share a letter are significantly different (p<0.05).

Muscle	Mean Maximum Muscle Force ± Standard Deviation	Group
Vastus Lateralis	1676.9 ± 221.3 N	A
Vastus Medialis	1012.3 ± 260.4 N	B
Gluteus Maximus	915.7 ± 304.6 N	B
Vastus Intermedius	729.1 ± 285.2 N	C
Biceps Femoris (long head)	539.9 ± 215.1 N	D
Rectus Femoris	437.0 ± 305.4 N	DE
Soleus	340.3 ± 309.6 N	E
Medial Gastrocnemius	98.1 ± 140.6 N	F

Table 2: Mean maximum muscle forces for each phase of STS transfer. The values represent the mean and standard deviation of the maximum muscle forces across the seven subjects for each phase of STS transfer, including both dominant and non-dominant legs. * indicates that the force generated by that muscle was significantly different than the other muscles examined in that phase (p<0.05).

	Mean Maximum Muscle Force ± Standard Deviation		
Muscle	**Phase 1**	**Phase 2**	**Phase 3**
Vastus Lateralis	1646.8 ± 269.9 N *	1764.3 ± 121.7 N*	1619.6 ± 232.5 N*
Vastus Medialis	944.6 ± 323.0 N	1085.4 ± 201.9 N	1006.9 ± 240.9 N
Gluteus Maximus	773.2 ± 202.3 N	1044.6 ± 270.8 N	929.2 ± 371.9 N
Vastus Intermedius	685.4 ± 311.3 N	809.4 ± 268.8 N	692.4 ± 277.4 N
Biceps Femoris (long head)	549.9 ± 248.2 N	579.0 ± 212.8 N	490.9 ± 186.4 N
Rectus Femoris	417.6 ± 181.9 N	458.9 ± 374.8 N	434.6 ± 346.0 N
Soleus	70.3 ± 100.4 N	359.7 ± 316.9 N	591.0 ± 212.6 N
Medial Gastrocnemius	73.0 ± 51.8 N	0.0 ± 0.0 N *	221.2 ± 179.6 N *

Inter-limb Differences

Inter-limb differences across individual phases of STS transfer had a large range of values for each of the muscles examined (Table 3). Some muscles, such as the soleus, also had negative percent inter-limb differences, implying that certain muscles generated a greater force in the non-dominant leg during the STS transfer. The only statistically significant inter-limb differences were for the gluteus maximus during Phases 1 and 2, semimembranosus during Phase 1, and the soleus during Phase 3. Inter-limb differences for the other muscles examined had large variability as some of the standard deviation values reached over 100%, such as those for the semimembranosus and semitendinosus muscles (Table 3).

Table 3: Percent differences between dominant and non-dominant leg maximum muscle forces. The values represent the mean and standard deviation of inter-limb differences across the seven subjects for each phase of STS transfer. * indicates a statistically significant difference ($p<0.05$).

	Percent difference [a]		
Muscle	**Phase 1**	**Phase 2**	**Phase 3**
Gluteus Maximus	21.53 ± 26.98*	29.72 ± 25.47*	26.48 ± 33.07
Gluteus Medius	39.93 ± 113.02	74.53 ± 149.86	10.52 ± 50.38
Rectus Femoris	8.81 ± 38.32	-32.77 ± 95.54	-8.70 ± 72.25
Vastus Medialis	13.69 ± 19.03	11.50 ± 21.03	9.71 ± 19.82
Vastus Lateralis	9.01 ± 13.22	4.99 ± 9.64	4.47 ± 7.91
Vastus Intermedius	11.74 ± 26.20	12.89 ± 29.03	12.08 ± 30.60
Semimembranosus	26.81 ± 78.88*	75.95 ± 116.93	38.47 ± 109.03
Semitendinosus	11.64 ± 112.77	63.46 ± 152.42	23.04 ± 99.89
Biceps Femoris (long head)	-21.22 ± 46.65	-20.16 ± 62.89	-7.32 ± 19.09
Medial Gastrocnemius	-15.58 ± 66.66	-24.19 ± 70.84	1.63 ± 50.57
Lateral Gastrocnemius	-6.54 ± 63.78	-24.21 ± 69.35	4.63 ± 53.09
Soleus	-18.86 ± 119.14	-47.93 ± 88.59	-21.23 ± 20.33*

[a] A negative percent difference indicates that the selected muscle generated a greater force in the non-dominant leg compared to the dominant leg.

Discussion

The purpose of this study was to examine individual muscle forces as well as identify inter-limb differences in maximum muscle forces per phase of STS transfer in a young, healthy population. To our knowledge, this is the first study to determine individual muscle forces during STS transfer using three-dimensional subject-specific musculoskeletal model simulations. The results confirmed our hypothesis that the vastus lateralis would produce large forces than the other muscles examined. Few inter-limb differences were also found to be statistically significant due to high variability between subjects.

Our EMG, kinetic, and kinematic results compared favorably to those reported by other studies. As in our findings, both Doorenbosch et al. (1994) and Mak et al. (2003) found that the knee torque generated during STS transfer was greater than the hip torque, then followed in magnitude by the ankle torque. The shapes of the joint torque curves from those studies were also consistent with our data. However, the magnitudes of the peak torque values in both studies were lower than what we reported which may be due to differences in the subject populations tested (Doorenbosch et al., 1994; Mak et al., 2003). Doorenbosch et al. tested a young, healthy population with more females than males (6 female, 2 male); Mak et at. tested an elderly population, with some considered healthy while others were diagnosed with idiopathic Parkinson's disease (6 healthy, 7 with Parkinson's); we tested more males than females in the young, healthy population (5 male, 2 female). In addition, subjects in Doorenbosch et al.'s study were asked to perform STS with their hands placed on their hips rather than crossed against their chest, which could have altered their STS transfer movement pattern and therefore, the joint torque values.

Two previous studies used two-dimensional models to determine muscle forces during STS transfer, yielding different results than our study most likely due to variations in modeling and calculations (Ellis et al., 1984; Yoshioka et al., 2012). Yoshioka et al. used a two-dimensional model with eight Hill-type muscles and determined that the quadriceps of one leg had a peak force of 2197 N while the gluteal muscles had a peak force of 413 N (Yoshioka et al., 2012) . In contrast, we used a three-dimensional model with 194 Hill-type muscles and determined that those muscle forces had higher peak values, 3715 N and 989 N, respectively. Ellis et al. calculated average muscle forces by combining results from four different extreme muscle activation conditions that were implemented on a simplified two-dimensional musculoskeletal model and found that in order to raise from a chair without the aid of arms, the average values of the maximum quadriceps, hamstrings, and gastrocnemius muscle forces for one leg were 5.50, 2.22, and 0.71 (times body weight), respectively,. For our study, we calculated average muscle forces by using static optimization implemented on a three-dimensional musculoskeletal model and found that the average force values for these muscles were 5.47, 0.94, 0.42 (times body weight), respectively.

Another surprising finding was the high variability in inter-limb differences among subjects from a young, healthy population. This could be due to the subjects not being able to use their arms to complete the task; they could have developed their own individual strategy to perform STS transfer, creating the high variances in the estimated muscle forces. We were also surprised that some of the muscles in the leg assigned as “non-dominant” generated a greater amount of force than those in the “dominant” leg for some or all of the phases of STS, as indicated by negative inter-limb differences reported in Table 4. This finding further illustrates the complex relationship between muscle forces and ground reaction forces as the muscles in the leg that generates the overall greatest ground reaction force do not always generate the greater force for each phase of the STS transfer.

There are several limitations that need to be considered when evaluating the results of this study. By not allowing subjects to use their arms, subjects are performing STS transfer in a way that is not typically done on a daily basis. However, patients are often asked to complete STS transfer with their arms crossed over their chest in a clinical setting when being tested to assess functional performance. Therefore, by staying true to clinical execution of STS transfer, our results will serve as a baseline and allow us to better analyze STS transfer of those diagnosed with various pathologies. In addition, because we only tested a young, healthy population the results from this study cannot be generalized for other populations, including the elderly or those with pathology, as they might yield different results.

Understanding individual muscle forces as well as symmetry of muscle forces between legs during STS transfer in healthy subjects is the first step to analyzing muscle function and weakness in subjects with conditions such as osteoarthritis. Through three-dimensional musculoskeletal modeling, we were able to demonstrate that while the vastus lateralis produces greater forces during STS transfer, the other vasti muscles as well as the gluteus maximus, soleus, biceps femoris long head, rectus femoris, and soleus muscles should also be considered necessary muscles as they assist in achieving an upright standing position. These results give us better insight as to what specific muscles are vital in maintaining a successful STS transfer, serving as a baseline for future evaluations of muscle function during STS transfer in various pathological populations.

Acknowledgements

This material is based upon work supported by the National Science Foundation

Graduate Research Fellowship Program under Grant No DGE-0822215. The author would also like to thank Mike McNally for his assistance in data processing.

Conflict of interest statement

The author does not have a conflict of interest regarding the contents of this manuscript.

References

Alexander, N.B., Gross, M.M., Medell, J.L., Hofmeyer, M.R., 2001. Effects of functional ability and training on chair-rise biomechanics in older adults. Journals of Gerontology Series a-Biological Sciences and Medical Sciences 56, 538-547.

Andriacchi, T.P., Alexander, E.J., Toney, M.K., Dyrby, C., Sum, J., 1998. A point cluster method for in vivo motion analysis: applied to a study of knee kinematics. Journal of Biomechanical Engineering 120, 743-749.

Arnold, E.M., Ward, S.R., Lieber, R.L., Delp, S.L., 2010. A model of the lower limb for analysis of human movement. Annals of Biomedical Engineering 38, 269-279.

Christophy, M., Faruk Senan, N.A., Lotz, J.C., O'Reilly, O.M., 2012. A musculoskeletal model for the lumbar spine. Biomechanics and Modeling in Mechanobiology 11, 19-34.

Crowninshield, R.D., 1978. Use of Optimization Techniques to Predict Muscle Forces. Journal of Biomechanical Engineering 100, 88-92.

Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E., Thelen, D.G., 2007. OpenSim: open-source software to create and analyze dynamic Simulations of movement. IEEE Transactions on Biomedical Engineering 54, 1940-1950.

Doorenbosch, C.A., Harlaar, J., Roebroeck, M.E., Lankhorst, G.J., 1994. Two strategies of transferring from sit-to-stand; the activation of monoarticular and biarticular muscles. Journal of Biomechanics 27, 1299-1307.

Ellis, M.I., Seedhom, B.B., Wright, V., 1984. Forces in the knee joint whilst rising from a seated position. Journal of Biomedical Engineering 6, 113-120.

Hanke, T.A., Pai, Y.C., Rogers, M.W., 1995. Reliability of measurements of body center-of-mass momentum during sit-to-stand in healthy adults. Physical Therapy 75, 105-113.

Holzbaur, K., Murray, W., Delp, S.L., 2005. A Model of the Upper Extremity for Simulating Musculoskeletal Surgery and Analyzing Neuromuscular Control. Annals of Biomedical Engineering 33, 829-840.

Holzbaur, K.R.S., Delp, S.L., Gold, G.E., Murray, W.M., 2007. Moment-generating capacity of upper limb muscles in healthy adults. Journal of Biomechanics 40, 2442-2449.

Hughes, M.A., Schenkman, M.L., 1996. Chair rise strategy in the functionally impaired elderly. Journal of Rehabilitation Research and Development 33.

Leon, J., Lair, T.J., National Medical Expenditure Survey (U.S.), 1990. Functional status of the noninstitutionalized elderly : estimates of ADL and IADL difficulties. Dept. of Health & Human Services, Public Health Service, Agency for Health Care Policy and Research, Rockville, MD (Parklawn Bldg., Room 18-12 20857).

Lord, S.R., Murray, S.M., Chapman, K., Munro, B., Tiedemann, A., 2002. Sit-to-stand performance depends on sensation, speed, balance, and psychological status in addition to strength in older people. The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences 57, 539-543.

Mak, M.K.Y., Levin, O., Mizrahi, J., Hui-Chan, C.W.Y., 2003. Joint torques during sit-to-stand in healthy subjects and people with Parkinson's disease. Clinical Biomechanics 18, 197.

Mehr, D.R., Fries, B.E.W., B. C., 1993. How different are VA nursing home residents? Journal of the American Geriatrics Society 41, 1095-1101.

Munton, J.S., Ellis, M.I., Wright, V., 1984. Use of Electromyography to Study Leg Muscle-Activity in Patients with Arthritis and in Normal Subjects during Rising from a Chair Annals of the Rheumatic Diseases 43, 63-65.

Rodrigues-de-Paula Goulart, F., Valls-Sole, J., 1999. Patterned electromyographic activity in the sit-to-stand movement. Clinical Neurophysiology 110, 1634-1640.

Scarborough, D.M., McGibbon, C.A., Krebs, D.E., 2007. Chair rise strategies in older adults with functional limitations. Journal of Rehabilitation Research and Development 44, 33-42.

Schenkman, M., Berger, R.A., Riley, P.O., Mann, R.W., Hodge, W.A., 1990. Whole-Body Movements during Rising to Standing from Sitting Physical Therapy 70, 638 - 648.

Schenkman, M., Riley, P.O., Pieper, C., 1996. Sit to stand from progressively lower seat heights - Alterations in angular velocity. Clinical Biomechanics 11, 153-158.

Schultz, A.B., Alexander, N.B., Ashtonmiller, J.A., 1992. Biomechanical Analyses of Rising from a Chair Journal of Biomechanics 25, 1383 - 1391.

Su, F.C., Lai, K.A., Hong, W.H., 1998. Rising from chair after total knee arthroplasty. Clinical Biomechanics 13.

Vasavada, A., Li, S., Delp, S.L., 1998. Influence of muscle morphometry and moment arms on the moment-generating capacity of human neck muscles. SPINE 23, 412-422.

Wretenberg, P., Arborelius, U.P., 1994. Power and Work Produced in Different Leg Muscle Groups When Rising from a Chair European Journal of Applied Physiology and Occupational Physiology 68, 413-417.

Yoshioka, S., Nagano, A., Hay, D.C., Fukashiro, S., 2012. The minimum required muscle force for a sit-to-stand task. Journal of Biomechanics 45, 699-705.